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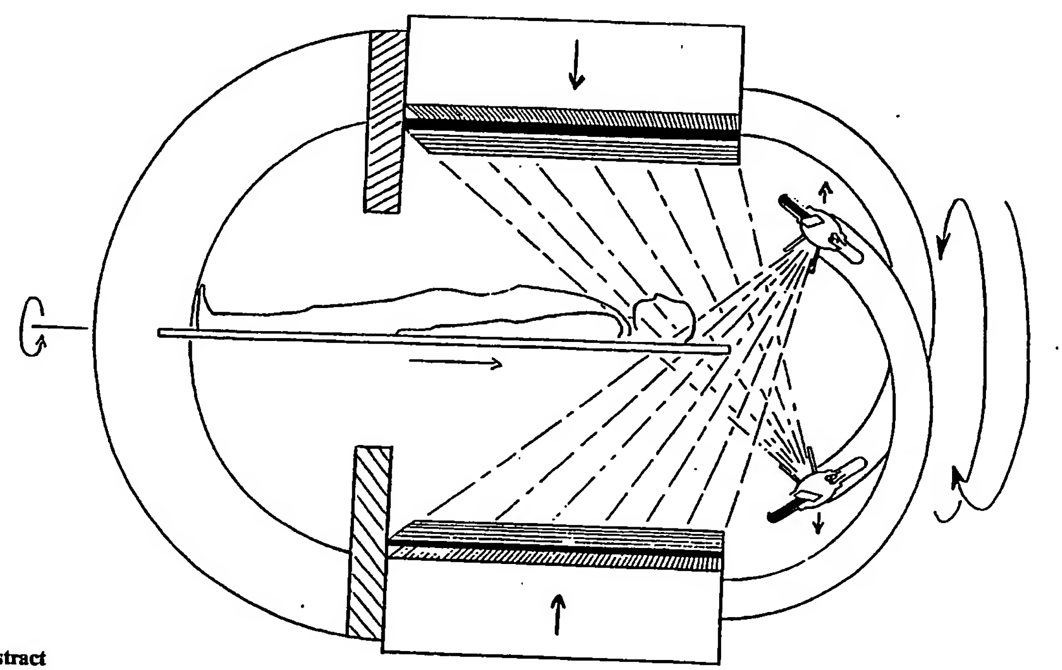

(57) Abstract

Acquisition of CT transmission data and omission data is achieved with the same scanner. Because the transmission and emission date sets are spatially registered, the map of attenuation coefficients described by the reconstructed transmission images can be used for generating a mathematically correct matrix for correction of sequentially or simultaneously aquired SPECT imaging data. The imaging devices can be gamma cameras (4).

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TRANSMISSION/EMISSION REGISTERED IMAGE (TERI) COMPUTED TOMOGRAPHY SCANNERS

ABSTRACT AND SUMMARY OF INVENTION

Reconstructed SPECT (Single Photon Emission Computed Tomography) images describe the intensity and location of sources of emitted photons in an attenuating medium, whereas transmission computed tomography (TCT) seeks to describe the distribution of different attenuation values within the attenuating medium. The photons emitted by inhaled, ingested or injected radionuclide sources in the body are attenuated by the amount and density of the attenuating material along a line between any emission source and the detector. The major problem in precise reconstruction of the SPECT image requires both the calculation of unknown activity in each part of the cross section as well as calculation of the unknown attenuation coefficients. The latter problem has been very poorly treated and is essentially unsolved by current methods which either make the (entirely incorrect) assumption of uniform gamma ray attenuation for all parts of the body or which grossly attempt to register CT images obtained by a different scanning device with different scanning geometry (and possibly different photon energy) for attenuation correction of a SPECT image. The present invention defines a system, a method, and several families of apparatus which make possible the

acquisition of transmission and emission data from the same scanner. There has, until the present invention, been no simple, safe and reliable method of creating transmission and emission images in exact registry on a multihead SPECT scanner.

The present invention defines a series of certain precise scanning geometries which consist of novel, unique and geometrically optimal arrangements, orientations and relationships between "fan-beam", "cone-beam" and "pyramid-beam" imaging collimators on the faces of gamma cameras (or other 2-dimensional x-ray imaging cameras or), and congruently positions similar fan beam, pyramid beam or cone beam collimated transmission sources which thereby permit the simultaneous or sequential acquisition of transmission CT images, for attenuation correction and anatomical correlation, in perfect spatial registration with the SPECT emission image.

Methods and apparatus are specified for the collimation of photons emitted from isotopic sheet sources into fan, cone, or pyramid shaped "transmission" beams, in a variety of different rotating and translating scanning geometries, using one, two and three gamma cameras. Also described are examples of isotopic point sources and X-ray tube point sources collimated to cone and pyramid transmission beam sources which can be directly substituted for the transmission cone or pyramid beam transmission sources collimated from isotopic sheet sources in other embodiments of the invention described herein. Although larger numbers of camera units are not illustrated, the basic principles and fundamental relationships described between the collimated transmission sources and the imaging collimators apply to any number or multiplicity of "opposing" cameras and transmission sources

arranged in rotating or translating arrays, and the invention is claimed to be trivially generalizable to arrays containing 4, 5, 6, 7, 8 and arbitrarily larger number of cameras.

In all of the implementations described herein, the precise positioning and geometric similarity of the diverging transmission fan beam, pyramid beam, or cone beam sources with respect to the opposing similarly shaped converging fields of view of gamma cameras "seen" through fan beam, pyramid beam or cone beam imaging collimators, overlays the focal lines or points of the transmission sources with the focal lines or points of the imaging collimators. This is a critical innovation common to all of this series of devices which makes possible the acquisition of spatially registered emission and transmission images from the same camera device. In the embodiments described which use gamma camera imaging devices, this arrangement of "gatable" transmission sources such as "switched x-ray tubes" or "moving sheet sources with or without" "shutters" permits the simultaneous or sequential acquisition of 2-dimensional transmission image data which, by virtue of its acquisition through the same imaging collimator that is used to collimate the emission photons, is inherently in perfect spatial registry with the acquired emission data. Because the geometries described herein inherently create spatial registration of the transmission and emission data sets, the map of attenuation coefficients described by the reconstructed transmission images is ideally suited to the purpose of generating a mathematically correct, precise, and exact matrix for attenuation correction of

sequentially or simultaneously acquired SPECT imaging data.

The use of multiple isotopic sheet or point sources in sequence, or the alternate constant current pulsing of the X-ray transmission source at different kilovoltage levels permits transmission image acquisition in a multiplicity of energy "windows" to facilitate "energy subtraction" which inherently increases the contrast resolution of the reconstructed image. The inherently superior energy resolution of gamma camera devices as compared to sensors typically used in transmission computed tomographic imagers, and the physical zoom feature described above, dictates that the theoretical and actual transmission CT performance of the devices described herein shall someday exceed that of all conventional transmission CT scanners existing today. Thus, even if used <u>ONLY</u> in transmission modes, these imaging devices have numerous advantages over conventional transmission CT scanners. Used in the combined TERI (Transmission/Emission Register Image) mode, these devices <u>shall</u> produce images of the body whose detail, quality speed of image acquisition and information content will far exceed that obtainable by any available conventional (non-TERI) S.P.E.C.T. (emission CT) scanner alone or any available conventional (non-TERI) transmission CT scanner alone.

Although not illustrated in complete detail, in all embodiments, the cameras and transmission sources are secured to rotating or translating "gantries" in precise registry, either by independent electronically controlled servos (such as those currently used by rotating multihead gamma cameras)

(e.g. figure 4) or by rigid. The former case requires that the servos maintain the relationship between the transmission sources and imaging collimators with great precision.

Providing that precise angular and distance spatial relationship between each transmission source and its opposing imaging collimator is maintained, by moving the convergent point of the transmission beams closer to the patient (and simultaneously moving the imaging collimator/gamma camera an equal distance further from the patient) all of the translating and rotating geometries described herein facilitate another unique, novel and dramatic improvement over existing methods of obtaining both emission and transmission data: i.e. Zoom acquisition.

BACKGROUND OF THE INVENTION

The development of transmission computed tomography (TCT) has permitted cross sectional images of the body to be generated based upon attenuation coefficients of tissue calculated from millions of measurements made by a rotating x-ray tube source transmitting a thin fan shaped beam of x-rays which are attenuated so they pass through body tissues to a ring array of detectors connected to a computer. A similar technology called single photon emission computed tomography (SPECT) has developed in the field of nuclear medicine imaging which utilizes a rotating gamma camera or a ring of rotating gamma cameras whose lead multihole collimators collimate the emitted photons to create an image on the face of the crystal of gamma ray emissions from organs and tissues containing radioisotopes which have been injected,

ingested or inhaled in the form of radiopharmaceuticals. In transmission computed tomography utilizing existing technology (i.e. prior to the present invention) only one "slice" can be obtained at a time, while current emission SPECT imaging devices permit a multiplicity of emission slices to be acquired in each scan. In some recent designs of SPECT scanners dedicated to the imaging of the brain (e.g. "SPRINT", "ASPECT"), the camera crystal is fixed (non-imaging) cylinder surrounded by photodetectors. Multiple angles of view in these devices are obtained by rotation or a multiholed or multislit collimator within the center of the annular crystal but outside (surrounding) the head, or body part to be scanned.

One of the biggest problems relating to this SPECT technology is that the emitted photons must traverse tissue and bone before arriving at the collimator and gamma camera detector and the path through this tissue and bone has an unknown attenuation coefficient for any given emitted photon. Schemes have been proposed to ascertain the attenuation coefficients of the tissue in the path of these emitted photons by obtaining a conventional CT scan of the body area of interest and to then attempt to generate registration of the conventional CT image with the SPECT image to utilize the computed attenuation image to correct the emission images. This approach suffers from the difficulty in obtaining exact registration of the computed tomography (attenuation) image obtained on one machine with the SPECT image obtained on a different machine.

Another method to calculate attenuation images has been to use planar sheet sources adjacent to parallel hole collimators opposite the patient from the gamma camera to cast a "gamma shadow" of the patent on

the camera. This procedure is cumbersome and exposes the technologists to unnecessary irradiation. It also produces a very poor "shadow" image due to the extreme "geometric unsharpness" effects associated with the parallel transmission collimator.

Other techniques have focused on simply ascertaining the body edge or boundary by evaluating scattered photons from the internalized radio-pharmaceutical using a separate energy "window" to collect an image of photons with energies below the "photopeak" of the isotope of interest scattered within the body. This scatter image is used in many schemes for attenuation correction to define the "body contour". However, in these schemes the body attenuation is quite incorrectly assumed to be some constant value, and areas outside the body are assumed to have zero attenuation, for purposes of calculating attenuation, resulting in a very poor corrections of the emission image. The very fact that planar x-ray and transmission CT images quantitatively and quantitatively differentiate body tissues of different attenuations demonstrates the fallacy of assuming a constant attenuation value in these calculations. The absurdity and inadequacy of the simple "body contour" attenuation averaging algorithms becomes obvious if one considers the several thousand-fold difference in the attenuation of emitted photons passing through lung and <u>between</u> the ribs as compared to those emitted photons whose path to the detector requires that they pass through a vertebrae body. This example describes a problem which can lead to misleading or perplexing artifacts in cardiac and other imaging modalities.

Among the problems seen in reconstructed SPECT images that occur as a result of the inadequacy of existing attenuation correction algorithms are "hot rim" artifacts and inaccurate data related to perceived asymmetrical organ uptake of radiopharmaceuticals. These problems present a dilemma for the clinician who is unable to determine whether a count reduction in a section of a reconstructed image is due to a genuine pathologically decreased uptake of a tracer or due to an artifact of uncorrected attenuation. Similar to the prior example is the frequent seen attention defect in the Thalium-201 cardiac imaging of large breasted woman where uncorrected attenuation from an overlying breast can mimic a pathological decrease in myocardial Thalium uptake worrisome for infraction or ischemia. Attenuation errors also decrease lesion contrast and thus the detectability of interior lesions, and in addition can cause volume deformation which makes it difficult to evaluate lesion sizes.

To "reconstruct" a set of cross sectional emission images in a standard SPECT scan, data is collected by rotating a gamma camera and collimator about a patient who has internally concentrated gamma emitting radiopharmaceuticals in the organ(s) of interest. The series of images obtained during rotation are processed by a computer which takes "line integrals" of these images and "back projects" them, after proper filtering, to generate cross sectional images. Algorithms exist to back project the images obtained from parallel hole, fan beam and cone beam collimation. Lead collimators serve to exclude any photon from the camera which does not enter parallel to the holes of the collimator. Generally, the

collimators used are attached to the face of the camera for imaging these incoming emitted photons. At the time of filing of the patent application for the present invention, the inventor is unable to locate any evidence of similar prior invention of methods or apparatus for simultaneous or sequential sequisition of transmission CT data on SPECT scanners for the purpose of optimizing attenuation correction.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENT AND EXPLANATION OF THE DRAWINGS

The present invention describes methods of implementation of these Transmission/Emission Registered Image (TERI) scans on both rotating and translating SPECT multicamera scanning systems. It has been shown by several authors that the placement of a Gadolinium 153 "point source" at the focal point of a converging collimator attached to a gamma camera can be utilized to generate a "bone density" image by subtraction of the attenuation images created by the two different photon energies emitted by Gd-153.

As yet, no invention has attempted to take advantage of this fact to generate computed tomographic transmission images, even though to do so would only 22require that this apparatus be attached to a rotating SPECT cameras equipped with reconstruction software designed for "cone beam" reconstruction of the transmission image (similar to that which has been developed for emission SPECT imaging by Ron Jaszczak of Duke University). The transmitted CT images could be registered in separate "energy windows" by the pulse height analyzer of the gamma camera to facilitate energy subtraction images, if two different isotope

sources are used, or if a dual energy isotope such as Gd-153 is used. Although the inventor has never tested these hypotheses, the physics described herein absolutely mandate the fact that a point source of radiation placed at the focal point of a converging collimator mounted to a rotating gamma camera could create a "transmission" CT data set which could be reconstructed into "transmission CT images". Since "transmission" scans produce images of tissue based upon attenuation, this data set could be utilized for attenuation correction.

This invention, consisting of an isotopic point source mounted at the converging "focal point" of an imaging collimator mounted on a single rotating gamma camera (or some other two dimensional X-ray-imaging camera or array) is claimed as the simplest implementation of a transmission/emission registered image (TERI) scanner. While it is possible to effect creation of a TERI scanner using only one transmission source and one gamma camera, the use of multiple rotating gamma cameras or multiple translating gamma cameras will greatly increase the speed of image acquisition and these multiplisymmetric multi-camera implementations of the invention are described as preferred embodiments. Although larger numbers of camera units are not illustrated, the basic principles and fundamental relationships described between the collimated transmission sources and the imaging collimators apply to any number of "opposing" cameras and transmission sources arranged in rotating or translating arrays, and the invention is claimed to be trivially generalizable to arrays containing 4, 5, 6, 7, 8 and arbitrarily larger number of cameras. Figure 7, for example, demonstrates 2-fold symmetry and 2

cameras. The extension to four-fold symmetry and 4 cameras is trial. It is claimed that all of the attached illustrations represent embodiments of the same invention.

THE USE OF X-RAY TUBES TO SUBSTITUTE FOR DUAL ENERGY ISOTOPIC POINT SOURCES

In the simple dual energy point source embodiment described above, the comparatively low photon flux of the isotopic point source could be improved upon if this source is replaced with the focal point of a low powered X-ray tube positioned at the focal point of the converging imaging collimator. Dual energy transmission in this case would preferentially require pulsing of the X-ray tube at two different voltages. Although not preferred, another method of generating a dual energy x-ray photon design could be effected by "chopping" of the X-ray beam with a rotating slotted metal (e.g., aluminum or copper) filter to create an X-ray photon beam of alternating average energies for energy subtraction imaging. In the designs described subsequently, this dual energy X-ray source could be substituted for the gamma ray transmission cone beam or pyramid beam sources by placing the X-ray tubes' focal points at the points of convergence of the opposing pyramid beam or cone beam imaging collimators (see Fig. 4, 7) (which in the descriptions to follow are congruent with the pinholes of a pinhole transmission collimators used to collimate photons from a gamma sheet source).

The use of X-ray tube sources instead of isotopic sheet sources (described below) in these "point focused" geometries of the TERI scanner has the distinct advantage of creating scanner designs which

have no internal radioactivity. The lack of monochromaticity of the X-ray tube source can be compensated for by aluminum, berylium niobium or copper filters and by the utilization of window settings of the gamma cameras which will count only the highest energy transmitted photons and discard all energies below a threshold value determined by the preselected energy of the X-ray pulse. This window setting could be synchronously gated to alternate values as the X-ray beam is pulsed alternately at two voltages (or as the slotted metal filter is rotated in the x-ray beam) in order to obtain alternating energy transmission beams to create images for energy subtraction and subsequent cone beam or pyramid beam back projection reconstruction. During acquisition of an emission (SPECT) scan with the above apparatus the X-ray tube could be unpowered or for "simultaneous" acquisitions it could be pulsed at each step of camera rotation or translation alter the emission data is acquired. A new design for an X-ray tube being developed by Quantum Optics Corporation may provide relatively monochromatic X-rays which would optimize this design and eliminate the need for beam filtering.

While the above "x-ray tube collimated point source" implementation would be useful for a triple head rotating TERI scanner (described below) using pyramidal imaging collimators, it could also be utilized in the translating TERI scanner described below if instead of asymmetric fan beam imaging collimators, asymmetric pyramid beam collimators were used, with convergence points congruent with the focal points of paired X-ray tubes. (See Fig. 4, 7. In these two figures the asymmetric pyramid beam collimators are made of crossed asymmetric radiographic

grids whose focal lines intersect orthogonally at the focal points of the x-ray tubes. This asymmetric pyramid beam imaging collimator could also be made with a "microcast" asymmetric pyramid focussing collimator (indicated by not actually illustrated in Figure 3).

All preferred embodiments of the present invention eliminate the image registration problem by using a set of collimator geometries and transmission sources which could be "attached" to a conventional single or multihead SPECT scanner, to convert it into a scanner capable of generating conventional (attenuation based) CT images in precise registration with the emission images. Because of the extreme precision required of the collimator relationships, it is unlikely that these inventions could be readily "retrofitted" to existing SPECT scanners.

All embodiments of this invention will also permit the superimposition of the emission (SPECT) computed image on the transmission (CT) image in color overlay so as to permit the exact identification of location of radioisotope tracers with respect to conventional anatomical boundaries defined on the computed tomography (transmission scan). The reconstructed transmission images as noted above also permit precise attenuation correction to be effected for all subsequent transmission images by virtue of the precisely registered overlay of the (attenuation based) transmission images.

Several implementations of this invention are described which utilize "line focusing" (fan beam) imaging collimators attached to square or rectangular head gamma cameras which are mounted in rotating or translating SPECT gantrys (Figures 1, 2, 3, 5, 6). In each design, the lines of convergence of the imaging

collimator are congruent with opposing lines of divergence of "transmission" collimators whose diverging faces are opposed to movable radioactive isotopic "sheet" sources. The sheet sources used for this purpose are movable by mechanical means from lead shielded boxes to the plane located behind the "transmission" (line focusing) collimators, so as to produce a fan beam which diverges exactly into the converging face of the opposing fan beam imaging collimator. (Figure 8) The use of servo-motors and lead shielded boxes to house the sheet sources has the benefit of preventing unnecessary irradiation of the technologists and it enables the rapid sequential acquisition of an emission data set following the transmission data set (simultaneous acquisition might also be possible in some cases).

Alternative embodiments are also described (but not fully illustrated at this time) which utilize symmetric or asymmetric "pyramid" beam (converging rectangular) or cone beam (converging circular) collimator geometries to effect the identical advantageous superposition of transmission and emission scans. The pyramidal collimator geometry could be utilized interchangeably in a TERI scanner to effect high resolution small parts scanning, utilizing the same moveable sheet sources behind pyramidal transmission "pinhole" collimators whose pinholes are congruent with the focal points of opposing pyramidal imaging multihole collimators mounted on the facing gamma cameras. The internal apical angle of the pyramidal transmission collimator should be the same as the convergence angle of the opposing multihole imaging collimator to effect minimal irradiation outside the periphery of the field of view of the imaging collimator. (The "end on" views seen in Figures 1 and

3 are very similar to the appearance of such pyramidal collimation. If items #1 in Figure 1 are replaced with pyramid shaped lead "pinhole" collimators, and the sheet sources are appropriately shielded, the exchange of items #3 for microcast "pyramid-beam" imaging collimators converts Figures 1 and 3 from "Fan beam" to "pyramid beam geometries."

As described above, it is possible to also effect transmission scanning by placement of the focal point of an X-ray tube in the position of the converging focal point of the imaging collimators (which would in the isotope sheet source implementations be congruent with the location of the pinhole of the transmission collimator).

In the case of the fan beam transmission (triangular multivaned) collimators, variation of collimator parameters such as the thickness and spacing of the triangular septae can be utilized to alter the thickness and spacing of the CT slices.

BRIEF DESCRIPTION OF THE DRAWING

FIG. 1 is a perspective view of a rotating transmission/emission registered image scanner of the present invention, using three sheet sources and three gamma cameras. Depending on the collimator geometries selected, Figures 1 and 2 could represent the "end on" views of either a rotating "Fan Beam" TERI scanner or a rotating "Pyramid Beam" TERI scanner.

FIG. 2 demonstrates "Zoom magnification mode of a rotating TERI scanner. In the lower illustration, note the insertion of lead attenuation "shims" in the "non-zoom" mode to narrow the spread of the fan transmission beam so as to eliminate scatter and reduce "noise".

FIG. 3 is a perspective view of a "translating" (TERI) dual head transmission/emission registered image apparatus. (Note that item 3 is not illustrated so a multihole microcast collimator.) (This device may also be implemented with 4 or more cameras.) If the 2-fold axial symmetry of this design is replaced by 4 fold, 6 fold or 8 fold symmetry, the same basic design provides for cameras that would surround the patient with square hexagonal or octagonal transaxial symmetry. Obviously, with such rotational symmetric transaxial symmetry, all of the above "translating" designs, could, with the proper design of a "cotating" gantry, also be configured to rotate so as to generate a larger number of different angular views for the purpose of more precise calculation of the back projection reconstructions of emission and transmission scans. A correlary of the fact that "higher order symmetry" generates a larger number of resistant axeous angular views is the intertwined fact that as the rotational symmetry increases from 2 fold towered n-fold, the angle through which the gantry must rotate to generate all possible views decreases proportionately. Thus, while a 2 fold symmetric device would have to rotate 180 degrees to generate all views, an 8-fold symmetry design would do so with only 45 degrees of rotation (and so on). This fact strongly favors the design of embodiments of the present invention with higher order rotational symmetry. Note that the drawing of this figure has technical errosion that the patient's body should be ideally centered above the axis of radial or rotational symmetry in all cases.

FIG. 4 demonstrates the replacement of the fan-beam imaging collimator with a pyramidal-shaped hinging collimator and replacementof the isotopial sheet source/multi-vaned triangular transmission collimator with assymetric pyramid-collimated x-ray point sources. Figure 4 is representative of a lateral cross-section of a 2 fold symmetric rotating and/or translating TERI scanner and is generalizeable to n-fold transaxial symmetry. The patient table's erroneous position notes in the prior drawing has been corrected here. Note that the x-ray tube sources, in addition to reducing potential undesired exposure to ionized "leakage" radiation, permit rapid switching from transmission to emission nodes and greatly increased magnification in the zoom mode, as compared to far beam designs, due to the two fold taxes of convergence of pyramid beams (as compared to far beams). This geometric magnification effect is also increased over that of the prior figure due to the decreased distance of the focal <u>points</u> of the x-ray tubes to the central axis (in comparison with the focal <u>lines</u> of the transmission collimators in the prior drawing, and the similarly <u>increased</u> distance of the imaging collimators and cameras from the patient.

FIG. 5 demonstrates in comparison to Figures 3 and 4, a low magnification (non-zoom) configuration of a 2 fold symmetric fan beam device which is analogous to the non-zoom "close hauled" collimator positions shown in the three-fold rotationally symmetric embodiment illustrated in the lower drawing of Figure 2. Again this cross-section is representative of 2 fold translating/and rotating TERI scanners. Figure 5 also provides an "exploded view" of the multi-vaned triangular fan beam transmission collimator, with 2 sliding radioisotopic sheet sources

(one of which is in imaging position below shich is seen a lead "shim", which reduces the dispersion angle of the transmission fan beam during "non-zoom" mode so as to bring the close-hauled beam angle into congruence with a shorter focal length fan beam imaging collimator. This is again analogous to the similar shims seen in the lower drawing of Figure 2. Note that the lower transmission sheet sources and shims are withdrawn into their protective lead boxes.

FIG. 6 demonstrates similar uses of these lead "shims" with "close-hauled" transmission fan beam collimators, but at increased magnification in comparison to the prior figure due to the longer focal length of the fan beam imaging collimators. Patient table height in this illustration is slightly above the ideal axis of rotational axial symmetry. The ability of this design (and the analogous pyramid beam designs to bring the transmission focal lines or points close to the patient, makes possible magnifications that would be physically impossible for a conventional CT scanner (or for SPECT emission scanners as they are currently being used). In Figure 7, 3-fold symmetric design is illustrated which (like Figure 4) utilizes pyramidally collimated X-ray point sources, and assymetric pyramid fan beam imaging collimators. Unlike Figure 4, this figure schematically demonstrates the rigid physical ________ of X-ray tube point sources on "connecting arms" which place their focal points in congruence with the focal points of the opposing pyramidal imaging collimators. Note that this relationship is maintained by the connecting arms (without complex servos), irrespective of decreasing zoom magnification factors (symbolized by the movement of the cameras and X-ray point sources in the direction of the vertical arrows). Note also that the patient

table position is in this Figure slightly above the ideal axis of rotational symmetry which should ideally be readjusted to overlap the axis mid-coronal plane of the body. Note also that the scanner geometry depicted in this drawing specifically demonstrates the potential motions of rotation and translation to obtain multiple angles of view for back-projection reconstruction. Since both motions increase the number of acquired angles, both may be utilized (either in sequence or simultaneously). Finally note that the vertical arrows in the center of the gamma cameras depicted in Figure 7 represent an additional axis of rotation of potential significant value, since 180 degree rotation about this axis would effectively double the number of angles sampled during a second rotational or translational scan.

The lower two illustrations of Figure 8 demonstrate additional details of the exploded view of the fan beam imaging collinator and "lead box-shielded" sliding sheet sources depicted in Figure 5, demonstrating the directions of motion of these sources, of a moveable lead shielding "shim" and of the leaded bottom of the box (through which access to these components are provided). Also noted in the lower left drawing is a lead "shutter" which could be momentarily slid or rotated over the focal slit of the triangular transmission fan beam collimator, so as to permit the "gating" or switching off of the transmission fan beam without moving the radioactive sheet source. This component makes possible the rapid alternation from transmission to emission modes, which could facilitate the acquisition of transmission images at multiple sequential angular positions of camera rotation before or after acquisition of the emission image of the same

angular view. Obviously an analogous lead "shutter" could be used to "gate" the transmission of a pyramid beam collimated isotopic transmission sheet source or point source. Finally, the top illustration of Figure 8 illustrates an alternative geometric embodiment of the sliding sheet sources and leaded box with respect to the traingular multivaned transmission collimator. This transmission geometry was illustrated previously in Figures 1 and 3, while Figures 2, 5 and 6 utilize the transmission geometry depicted in the lower two illustrations of Figure 8.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS AND DRAWING

Although the present invention can be implemented with a single gamma camera (as noted above), preferred embodiments all have "higher order" symmetrics composed of a multiplicity of cameras and transmission sources. The preferred rotating scanner geometry for three-fold symmetry is an improvement upon and modification of the three head SPECT camera designs currently being manufactured by companies such as Trionics (TRIAD) and Ohio Nuclear (PRISM), although, as it will be further described below, it is also possible to create Transmission/Emission Registered Image (TERI) scanners with a non-rotating, translational linear scanning approach, as might be effected by modification of any dual head, linear translating emission body scanner (such as the EWF scanner made by Siemens), using the modifications of such linear scanners previously described by the inventor in confidential disclosures and illustrated in this disclosure in Figures 3, 4, 5 and 6).

The general use of fan beam imaging collimators in single, double or triple headed rotating

SPECT scanners has been avoided by most current manufacturers because of the perceived difficulty in attaining a precision fan beam collimator design from which reliable back projections can be derived. Recent advances in microcast collimators using technology developed by the Nuclear Fields Company of the Netherlands has made possible the creation of greater precision in the alignment of holes to converge upon a line focus. As described above, and illustrated in Figures 3, 4, 5, 6 and 7, it is also possible to create effective and precise fan and pyramid beam imaging collimators by the placement of two orthogonally oriented radiographic image grids over the face of the camera. When one of these grids is "converging" and the other parallel, this results in a "Fan beam". When both are converging it results in a "Pyramid beam". The use of converging fan beam collimation in rotating SPECT cameras permits the SPECT scanner to take advantage of the greater extrinsic resolution potential of the line focusing collimator for imaging (as compared to "parallel hole" collimators). Since the "holes" of such an imaging collimator converge upon a line, photons coming from an opposing identical "transmission" collimator attached to a sheet source of a radioisotope could be made to "flood" the field simply by precise alignment of the focusing "lines" of the transmission and imaging collimators. (Figure 1, 2, 3, 6).

Photons from such a "transmission" fan beam collimator can be directed through the line focus to diverge through a patient and register an attenuation image on the face of the opposing gamma camera after passing through the line focus imaging collimator (Figures 1, 2, 3, 6). In practice, it is possible to make the "transmission" collimator smaller than the

"imaging" collimator, if the angle of the fan beam is the same. It is also possible to use the Triangular Multivaned Transmitting Collimator design (of Figure 8) in which triangular lead septa separated by gamma ray "transparent" spacers collimate photons in one axis while a "slit" at the apex of the multiple triangular vanes defined by the edges of angled lead side plates collimates photons in the orthogonal axis. This design, unique to the present invention, is geometrically superior to multihole cast collimators for the purpose of effecting a highly collimated fan beam. Since source collimation is a crucial determinate of transmission image sharpness, it is asserted that this design is the optimal fan beam transmission collimator technology, and that furthermore it will effect more even distribution and higher flux than would be obtained using a multiple angled hole design. The slit at the apex of the triangular septae would be congruent with this line focus of the imaging collimator to effect registration of transmitted photons with the focus line for emission and transmission reconstruction of the fan beam images. By making the width of the "slit" adjustable, and by making the gamma ray transparent spacers out of a compressable material (such as neoprene) this collimator can be modified to have the unique property of independently variable "resolution", and "sensitivity" both parallel and perpendicular to the imaging slit.

The fact that the triple head scanner designs presently in existence have not utilized fan pyramid or cone beam geometry for transmission images to date is because the manufacturers have so far either failed to recognize or to capitalize upon the advantages of fan beam or converging collimator geometries even for

emission imaging (not to mention their potential application for transmission imaging). Properly designed microcast line focusing collimators, cone beam and pyramid beam collimators offer superior resolution and sensitivity for SPECT imaging. The collimator invention of Figure 8, and its variable resolution modification described above would also offer superior resolution as "imaging" collimators which would be manifest as a smaller "full width at half maximum" (FWHM) system resolution. No company to date has demonstrated a transmission CT image obtained from any rotating gamma camera, utilizing fan beam, pyramid beam or cone beam reconstruction algorithms, or a scanner designed to optimize the registration of transmission and emission images. No collimator to date has ever purported to offer variable resolution and sensitivity in both orthogonal dimensions (although, the variable slit width seen in the "SPRINT" annular scanner at the University of Michigan designed by Dr. Leslie Rogers does offer these variables in <u>one</u> of the two orthogonal planes of collimation. The application of compressible gamma ray "transparent" spacers to the spaces between the lead "axial" collimator rings in Dr. Rogers' scanner would provide variable sensitivity and resolution in the axial dimension as well and is claimed here as a parallel invention for creation of similar collimator adjustability in an "annular" scanner using rotating slit/parallel ring collimation (Figure 9).

<u>ROTATING TERI SCANNERS</u>

The rotating triple head design illustrated in Figures 1 and 2, is geometrically ideal for rapid Transmission/Emission Registered Imaging (TERI) scanning because the apex of the triangle created by

two adjacent square or rectangular gamma cameras is in the ideal location for the placement of a fan beam (or pyramid beam) transmission source collimator (pyramid beam modification to be described below) (See Figure 1). The ideal fan beam transmission source collimators are as described above, illustrated in Figure 8. The preferred embodiment of the rotating TERI scanner utilizes three rectangular gamma cameras arranged as in Figure 1 on a rotating gantry with either fan beam (as illustrated) or pyramid beam (interchangeable) imaging collimators whose line focuses (or focal points) are congruent with the line focuses or convergence points of opposing line focusing fan beam or pyramid beam "transmission" collimators, and whose angles of convergence are identical with those of the imaging collimators. This geometry maximizes the use of the gamma camera crystals both for transmission and emission scanning. The use of interchangeable "transmission" and "imaging" collimators permits the cameras to be separated (moved outward) to accommodate larger patients or (body parts) since this only requires the selection of collimators with longer "focal lengths" and adjustment of the cameras and transmission collimators positions with servos already standard in the rotating gantrys of existing triple head scanners. If the cameras are outward while the opposed transmission sources are moved inward an equal distance this design permits continuously variable geometric "zoom" magnification as illustrated in Figure 2.

The rectangular gamma cameras are preferred over circular designs because there is no lost space between them in the triple quadruple, and higher multiple head rotating designs, but the rotating TERI scanner concept is equally applicable to one or more

rotating circular cameras using converging collimators and cone beam reconstruction. The initially described embodiment using a gadolinium point source at the focal point of a cone beam converging imaging collimator is suboptimal because the Gadolininium point source (or any isotopic "point" source) will have significantly lower flux than a corresponding "sheet" source, and because the use of a dual energy isotope will invariably introduce some scatter photons from the higher energy emission into the "imaging window" for the lower energy emission (which would ideally be exclusively windowing only unscattered events to produce a transmission image). This problem is eliminated in the present invention through the use of six possible "dual energy" photon sources.

X-RAY POINT SOURCES

The first method of producing "dual energy" photons involves the placement of the focal point of an X-ray tube at the convergence point of the imaging collimator, with a small transmission collimator to restrict the divergence of the X-ray beam to the face of the opposing converging (pyramid beam or cone beam) imaging collimator. This X-ray tube is pulsed at two different voltages sequentially, to produce two different maximum photon energies for imaging. A thin Niobium, aluminum berylium or copper filter is used to "harden the beam" by screening out undesirable low energy photons and the pulse height analyzer circuits of the gamma camera are utilized as an electronic filter to reject all photons below a preset threshold (within say 10% of the peak KV of the X-ray tube voltage pulse). The X-ray tube could be alternately pulsed with two different voltages to produce diverging

X-ray photon beams of two separate peak energies which could be registered in image memory locations selected by windows using two separate pulse height analyzers which are set to these separate energy peaks. The pulse height analyzers would be alternately gated on to register the received image corresponding to the peak energy of the photons being generated at each X-ray tube voltage, to produce one image for each pulse. (Figure 10)

Rapid alternation of the X-ray energy spectrum could also be effected by an X-ray of constant kilovoltage which is interrupted by a rotating slotted metal filter which will sequentially harden (or soften) the X-ray beam by alternately removing the softer (more easily attenuated X-rays). The application of dual energy X-ray pulsing to the cone or pyramid X-ray beam of the present invention will not only make it possible to acquire a bone densitometry scan in a fraction of the time required by other bone density scanners but it will also permit this scan to be acquired in three dimensional cross sectional CT images if desired. The dual voltage pulsed X-ray source described above are preferable to the use of a rotating slotted metal filter for the present invention. The combination of two peak X-ray voltages and two pulse height analyzers set as "high pass electronic filters" to image the high energy photons will effectively create transmission images of relatively monochromatic X-ray photon energies which will more effectively simulate isotopic sources. The use of such windowed dual energy transmission imaging will permit precise calculation of attenuation coefficients by the computer and expansion

of the dynamic range of the CT "grey scale", improving the potential "contrast resolution" of the images. These can be used for precise attenuation correction or for radiographic correlation with emission images.

A new design for an "X ray laser" tube being developed by Quantum Optics Corporation may provide relatively monochromatic X-rays which would optimize this design. The transmission images could be obtained and recorded into image memory locations which could then be superimposed upon SPECT (emission) images obtained subsequently in the identical locations within the patient and which are stored in corresponding emission memory locations. During the acquisition of the emission scans, the X-ray tubes would be unpowered.

The use of pulsed X-ray tubes or gamma point sources and converging collimators to obtain TERI scans using pyramid beam or cone beam reconstruction algorithms is an application which will provide magnification TCT and emission imaging of small body parts of the cameras are moved outward and longer focus collimators are used. It is also possible to generate the dual energy diverging photon beam from three other possible transmission source geometries which utilize radioactive sheet sources collimated by rectangular or circular pinhole collimators (for pyramid beam or cone beam reconstructions) or by triangular multivaned slit collimators (for fan beam reconstruction).

The design illustrated in Figures 1 and 2 demonstrates the use of fan beam collimators (although in planar view this is identical to a drawing of the appearance of the rotating design using pyramid beam collimators). In both cases the lines of convergence of the imaging collimator are congruent with the opposing lines of divergence of transmission collimators whose diverging faces are opposed to moveable

radioactive isotopic "sheet" sources. The line focusing transmission collimators used in the "fan beam" acquisition mode of a rotating or translating TERI scanner may be of conventional multihole microcast design, but would preferably be designed as a multi-vaned triangular slit collimator with the slit at the apex congruent with the line focus of the opposing imaging fan beam collimator. Details of this triangular multivaned slit collimator are described above.

TRANSLATING TERI SCANNER

The use of line focusing collimators on rectangular translating gamma cameras to create SPECT images includes a method of indexing the images obtained from a pair of opposed translating rectangular gamma cameras attached to line focusing (fan-beam) collimators to create data sets which may be reconstructed into whole body emission images by SPECT reconstruction software. This invention also includes a design for a hardware interface data buffer for sorting the digitized gamma camera data into a multiplicity of different angular view memories, each of which represent a single angle of view, as projected through a single line of angled holes from the line focusing collimator onto a single thin rectangular section of the gamma camera crystal parallel to the line focus. During translation of the gamma cameras and collimators, each of these multiple thin slices of the crystal acquire a different angle view of emitted photons from the patient. The individual angle view array memories are each connected to a counter which will simultaneously increment the input of each memory

raster to a new line as the translating gamma camera pair advances (translates). Each angle view memory will acquire a raster which is effectively a whole body angled view, which can be reconstructed by SPECT backprojection methods into cross sectional tomographic images of the whole body (as such angled views are reconstructed by conventional gamma cameras). One embodiment of this design used an asymmetrical fan beam collimator on each face of the rectangular translating gamma camera and utilizes the angle view memory array described in block diagram in Figure 11.

In the optimal embodiment of the translating TERI scanner invention, described in Figures 3, 5, and 6, an asymmetrical fan beam (line focusing) imaging collimator is attached to each of two opposed rectangular faced gamma cameras with the line focus of each collimator angled off center, so as to converge lateral to the edge of the opposing imaging collimator on a line congruent with the slit at the apex of a multivaned triangular "transmission" collimator, behind which a servo movable isotopic sheet source of radiation can be automatically advanced from a shielded lead box. This sheet transmission source can be utilized in a manner analogous to that described above for the rotating TERI scanner, to acquire a whole body transmission CT data set that is in perfect geometric registry with emission data sets which can be subsequently acquired from the same translating rectangular gamma camera scanner (with line focusing collimators) described above. Analogous to the advantages cited for a <u>Rotating TERI Scanner</u>, this <u>Translating TERI Scanner</u> geometry permits both perfect attenuation correction by the acquisition of a "registered" transmission image as well as permitting

the superposition of "color overlays" of the emission and transmission images in addition to the potential for image subtraction, as described for the Rotating Teri Scanner. The design cited above may be used to generate even higher resolution images of selected body areas by rotating the patient or both collimator sets (or cameras and collimator sets) 180° so as to acquire double angles of view of the patient on a second pass scan of the same area either in transmission or emission modes or both (Figure 7). This would require a second set of "angle view" memories (Figure 11), for transmission and emission data and would double the number of views from which the TCT or SPECT reconstruction would be acquired.

FAN BEAM TRANSMISSION COLLIMATORS

The preferred embodiment of the fan beam transmission collimator (See Fig. 8) is a design consisting of a stack of triangular lead plates separated by alternating triangular glass or plastic (or compressible gamma transparent) spacers, with two rectangular lead side plates covering two of the triangular faces of the multivaned collimator meeting at the apex where these two faces converge as an apical slit. The third face of the triangular multivaned transmission collimator is opposed to a radioisotopic sheet source (instead of a leaded plate). Photons emitted by the radioisotopic sheet source in the direction of the apex and parallel to the adjacent lead plates will traverse the gamma transparent spacing material and escape through the apex between the lead face plates as a multiplicity of finely collimated parallel fan beams.

In both the rotating and the translating TERI scanner designs, the triangular septae slit edge transmission collimator can be utilized to generate transmission computed tomographic slices whose spacing, thickness and acquisition times can be varied by alteration of the spacing and thickness of the triangular gamma transparent septae. For example, the use of transmission collimators with thick septae which are closely spaced and with narrow apical slits would permit generation of thin, high resolution transmission CT (TCT) slices whose acquisition could be made faster, and to have even higher spatial resolution through the use of high activity radioactive sheet transmission sources and slow gantry rotation (to increase imaging "statistics"). By using a compressible gamma transparent material such as the oprene for foam rubber, the application of pressure by a "screw type" adjustment at each triangular lead end plate allows all of these parameters to be "continuously" variable. Likewise the adjustment of the slit to variable widths will change these parameters in the orthogonal direction. To effect the very sharpest possible "knife edge" slit, the apex of this high-resolution transmission imaging collimator could be made of depleted uranium (a concept the inventor credits to Dr. Leslie Rogers of the University of Michigan). To acquire a complete transmission image set, the patient could be incrementally advanced into the rotating TERI scanner during sequential rotations of a transmission scan to acquire the cross sectional images between the septal spaces of the first scan. Thicker lead septae could be used to collimate photons of sheet sources with relatively higher energies.

For purposes of rapid transmission CT image acquisition to effect registered attenuation correction,

thinner lead septa with wider gamma transparent spacing might be used to generate transmission CT images of lesser resolution. It is also possible to use thin septae with narrow spacing to increase the flux of the isotope-generated fan beam. This will permit the simultaneous acquisition of contiguous transmission CT images, and would be preferable because of the concurrent decrease in the time needed to acquire an image.

RADIOISOTOPE SHEET SOURCES

As shown in FIGS. 1 and 3, behind each transmission collimator (1) (which may be either the conventional cast lead variety or preferentially the triangular multivaned variety described above) is a movable isotopic sheet source (2) which will supply photons for transmission imaging when it is physically moved from its lead shielded box (5) into the plane behind the transmission collimator (1). There are preferably at least two different sheet sources used to generate the transmission image data sets which can be individually slid into place by automatic advance of the desired sheet source from a slotted lead box (5) to a position overlying the rear face of the transmission collimator (1). Since it is necessary to obtain two separate energy images for the energy subtraction needed to create a full density spectrum CT image, the isotope sheet sources used must be of at least two different gamma energies. These preferably have long half lives and photon energies in a range which permit differential absorption by a different body tissues. The preferred isotopes for this purpose include Gold-198, Cobalt-57 and Iodone-125, although other

isotopes could be used. The use of at least two separate moveable sheet sources to create two different energy level transmission images in the preferred embodiment has the distinct advantage of creating images which can be "windowed" to the narrow energy peak of each separate isotope to create an "energy subtraction" image which permits quantitative measurements of attenuation such as would be needed to measure bone density. The rotating and translating TERI scanner geometries described herein would permit these sheet sources to be used to simultaneously acquire transmission images on all of the gamma cameras. The spacing of the transmission computed tomographic images and their "thickness" could be varied by changing the thickness and the spacing between the triangle septae of the imaging collimator.

A similar effect using a dual energy emitting isotope such as Gadolimium-153 has the disadvantage of "spill down" of degraded scattered higher energy photons into the lower energy "window" of the gamma camera which will degrade the lower energy transmission image, and which is unnecessary if separate images of monoenergetic sheet sources are obtained.

Given two separate photon energies from two separate sheet sources which are advanced into the rotating transmission collimator during two sequential rotations of the triple head scanner, there will be two different "energy attenuation, "image sets generated by photons traversing the tissue within the SPECT scanner. Given the basic equation for attenuation of photons $I=I_{\theta}(I-e\mu^{x})$

where -->I is the intensity of the incident photon beam and
where -->μ is the unknown attenuation coefficient and -->x is the distance traversed,

the SPECT scanner's computer can be programmed to generate an attenuation map of cross sectional images by solving multiple simultaneous equations for μ in the same way as is done by a conventional CT scanner. The current invention describes a geometrical arrangement of components and a method whereby the gamma camera replaces the conventional CT scanner's sensors; and the collimated transmission fan beam from the dual energy isotope sheet source replaces the x-ray tube to create a CT scanning device which will generate many slices simultaneously (during transmission scans). Conventional CT scanners generate only one slice at a time. Furthermore, these CT transmission slices can be aligned precisely to generate a Transmission/Emission Registered Image (TERI scan) which permits precise use of the transmission images to create an attenuation map for attenuation correction of the emission image. This will also permit the localization of the emission image within the anatomical cross sections of the transmission CT (attenuation based) images, for the purpose of radiographic correlative imaging and subtractions.

The use of servomotor controlled moveable radioactive sheet sources will minimize any potential unnecessary irradiation of the technologists and the patient during the TERI scan, however these could also be moved manually with levers and or gears.

<u>ELECTRONIC SIGNAL PROCESSING</u>

Figure 11 shows a block diagram for a computer interface buffer which includes a FIFO memory buffer controlled by the gamma camera pulse height analyzer and two microprocessor control circuits.

The microprocessor control circuits distribute data from the gamma cameras into random access (angle-

view-array) memories for the purpose of optimizing subsequent image processing.

The multi-energy, random access, angle-view-memory array consists of 3 sets of 256 RAM chips, each of which contains an array of 4096 by 256 memory locations, which are each eight bits deep. The choice of which of the three arrays into which an incoming pulse will be stored is based upon the energy of that pulse as determined by the gamma cameras' pulse height analyzers. These activate an array selector gate which, through a pulse distribution microprocessor, instructs the imaging microprocessor to sort the incoming pulses based upon their energies into one of the three sets of 256 RAM chips. This initial sorting of the pulses by energies permits three separate spatially registered images to be stored from three different photon energies.

For a given set of 256 RAM Angle View Memory Chips, (i.e. for a given photon energy) the incoming pulses are sorted into one of these 256 chips on the basis of an 8 bit digitization of the "Y" pulse of the gamma camera which activates an 8 bit-binary to one of 256 selector circuit to enable one of the 256 RAM chips to receive the incoming address specified by an 8 bit by 12 bit address clocked from FIFO memory number 1 into the common 8 x 12 bit address lines. The Random Access Memory location thus chosen dumps its contents onto an 8 bit output line which is loaded into an 8 bit parallel load counter. The counter is incremented by one from instructions received from a timing and control circuit and the contents of the counter are then reloaded into the chosen memory location (which now contains (n + 1) counts. This apparatus stores the incoming selected events in different RAM chips based

upon the "Y" location of the pulse (which with fan beam collimation will be angle dependant). Within any given 4096 x 256 RAM angle view memory array, the position along the 256 dimension is determined by the digitization of the "X" signal while the position along the 4096 dimension will be determined by the product or sum of the digitization of the "Y" signal and the analog gantry position signal "P" (created by an arithmetic logic unit).

The digitized outputs of the gamma camera and the arithmetic logic unit are all buffered by a 42 bit by 1536 byte FIFO memory which stores digitized pulses and clocks them into the angle array memory buffer based upon the instructions from a timing and control circuit activated by the imaging microprocessor. A separate timing and control circuit controls the downloading of the array memories into the inputs of an external computer or computers for the purpose of image reconstruction.

The angle array memory buffer will store images from a translating or rotating gamma camera equipped with a line focussing collimator in a series of memories which each have only one angle of view of the whole patient. This permits conventional SPECT reconstruction algorithms design for parallel hole rotating collimation to be applied to data acquired from a translating or rotating fan beam collimator camera.

I CLAIM:

1. A method of creating an attenuation-corrected single photon emission computer tomography image (SPECT) image comprising the steps of:

(a) acquiring a transmission CT scan from a patient in a specified registration to a gamma camera sensor by a converging "fan," "cone" or "pyramid" beam imaging collimator whose focal line or point is congruent with the local line or point of a similar fan, cone, or pyramid beam X-ray or gamma ray transmission source;

(b) while maintaining said registration and imaging geometry between the patient and the gamma camera, acquiring an emission scan;

(c) correcting the emission scan diametrically to compensate for parallax are aberration so as to clarify acquired image; and

(d) correcting the emission scan diametrically to account for the calculated attenuation determined by prior calibrated error (see A).

2. A method of creating CT transmission images using one or more gamma camera sensors comprising the steps of:

(a) providing one or more sources of monochromatic photons, and collimating said one or more sources of photons to provide diverging beams which pass through a patient to one or more gamma cameras;

(b) providing a multihole, converging collimator on the face of each of said one or more gamma cameras, having a focus convergent with the focus of divergence of the respective one or more diverging beams of photons, and

(c) translating or rotating said one or more photon transmission sources, converging imaging collimators and gamma cameras with respect to said patient (while maintaining registration described in claim 1) to acquire multiple different angled planar views of said one or more beams as attenuated by said patient;

(d) storing data from said planar views in a computer image memory and;

(e) reconstructing three dimensional cross sectional images from said stored planar views.

3. A method of arranging and aligning a set of "transmission" collimators with respect to the position of imaging collimators with specific precise geometries to create collimated, shielded, transmission gamma, photon beams whose dimensions are congruent with the imaging fields of opposed identical imaging collimators attached to rotating gamma cameras for the purpose of generating transmission images for computerized tomographic reconstruction.

4. These collimator geometries are claimed as optimal arrangements for the production of transmission images which will be in perfect congruency with emission images obtained from SPECT gamma cameras using "fan beam," "pyramid beam" or "cone beam" reconstruction algorithms.

5. The design of moveable gamma emitting isotopic planar sheet sources attached to servo-motors or levers which will place one such source at a time behind the transmission collimator during a transmission CT scan to generate a collimated photon beam. These sheet sources to be automatically returned to a lead shielded box either manually or by remote control so as to spare unnecessary exposure of the technologist and patient and to facilitate transmission imaging with sheet sources of differing photon energies for energy subtraction.

6. The design of multiple moveable transmission sheet sources in the present invention is also claimed to produce images which permit quantitative measurement of attenuation such as would be useful for bone densitometry.

7. The design of these scanner geometries which permit the optimal registration of emission and transmission images is also claimed to facilitate color overlay projections and subtraction of radioactive uptake areas in the emission scan in comparison with definable anatomic margins on the transmission CT image. This method will optimize radiographic correlative imaging and image subtraction. No SPECT scanner currently in existence is designed to permit rapid superposition of transmission and emission computed tomographic scans for this type of comparison.

8. The Transmission/Emission Registered Image scanner (TERI) geometries described herein are also claimed to be uniquely efficient and an optimal design to calculate attenuation coefficients prior to re-

construction of emission SPECT images for purposes of attenuation correction.

9. The modification of a Gadolinium-153 point source bone densitometry apparatus with converging collimator attached to a gamma camera to produce transmission CT images. This modification being the attachment of said apparatus to a rotating SPECT camera with cone or pyramid beam reconstruction algorithm and dual window acquisition to collect transmission CT data for "cone beam" or pyramid beam reconstruction. This is again a claim applying known spectographic principals which, although as yet unproved, should permit acquisition of registered transmission/emission images as described above. This same spectography could also be used with two different isotope point sources in sequential transmission images as described above.

10. A method of generating a high resolution, modulated image data set from the outputs of a pair of translating gamma cameras attached to line focusing collimators. The digitized X and Y outputs of said gamma cameras in this method are directed to one of a multiplicity of "angle view array memories" which are selected by gating circuitry to acquire an image from thin multiple "slices" of the crystal each of which "sees" a different angle of the body during translation of the gamma camera pair. The circuitry selects an angle view memory based upon the "Y" position of an event and stores points within the angle view memory array for each valid (windowed) event recorded in the gamma camera crystal. In addition to the X and Y positions of each event along a given line within the array memory, the line in which any given event is recorded will be determined by the instantaneous gamma

camera position when that event is recorded. This method is hypothetically applicable to the use of a symmetrical fan beam collimator to create SPECT images (as was described in the original disclosure in 1985), however, the preferred embodiment would utilize asymmetrical fan beam imaging collimators for the purpose of facilitating transmission/emission registered image scans.

11. The design of sliding sheet sources of radioisotopes moved by servomotors from shielded boxes to active positions behind transmitting collimators which generate imaging "beams" to create transmission gamma photon images of a patient while minimizing the dose to technologists. Multiple sheet sources of different isotopes stored in such lead shielded boxes could be used to generate transmission images which assess the attenuation of different photons of interest so as to create attenuation maps for attenuation correction that relate to isotopes of interest. A Technesium 99m sheet source, or a sheet source of any other short lived isotope could be created for this purpose by injecting the isotope of interest and mixing it in a flat plexiglass box filled otherwise with water and sized to fit within the sheet source, servoracks behind the transmission collimator. Other solid isotope sheet sources of interest specifically claimed as important for full implementation of this invention would include Gold 198, Iodine 125, iodine 128 and Cobalt 57.

12. The design of a unilaterally congruent triangular transmission photon generator whose parameters may be varied to change the scope or intensity of the beam. The high-resolution transmission collimators may require knife edge slits

of depleted uranium to effect the very sharpest possible line focus of transmission to reduce "geometric unsharpness".

13. The design of interchangeable transmission collimators to effect different transmission beam parameters for lower resolution, high flux transmission imaging needed for attenuation correction. These collimators would have thinner septae and possibly wider spaces.

14. The application of interchangeable "pyramid beam" or "cone beam" collimators to replace the fan beam collimators described in the above Rotating TERI Scanner for generation of high resolution small parts TERI images from rotating rectangular or circular cameras fitted with pyramidal or cone beam reconstruction algorithms and software. These inventions are otherwise similar to the considerations described for the fan beam collimators. In these embodiments, however, a conical or pyramid pinhole transmission collimator could be used to collimate the photons from the transmission sheet source to a point congruent with the point of convergence of the opposing conical or pyramid beam imaging collimator. To create the sharpest possible pinhole transmission (and to minimize "geometric distortion") the pinhole could hypothetically be made of depleted uranium. These embodiments of the TERI Scanner with proper reconstruction software will permit high resolution small parts scanning and zoom magnification.

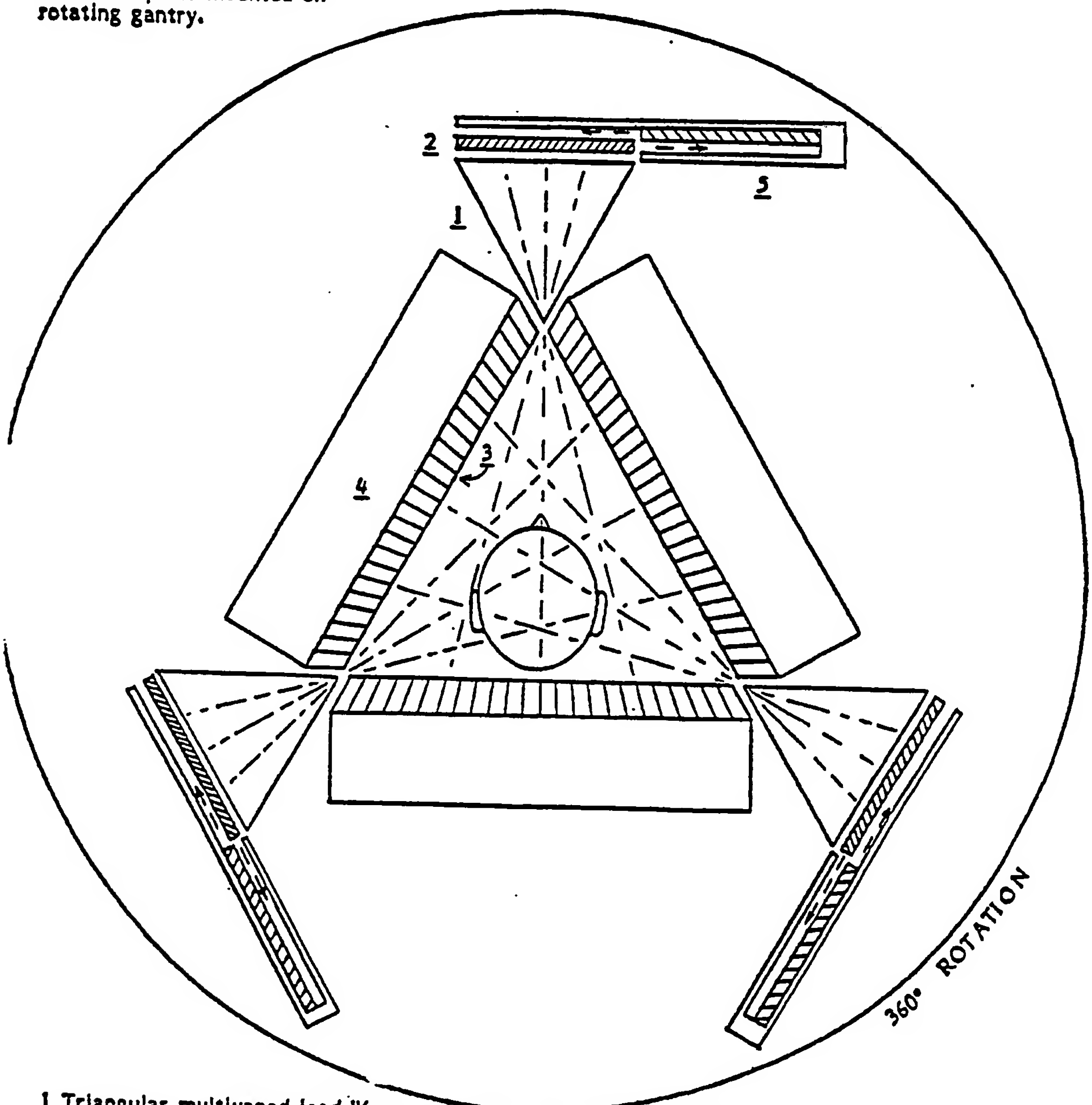
15. The use of X-ray tubes as point transmission sources in combination with gamma cameras as detectors to create transmission images for transmission CT

reconstruction using converging (cone beam or pyramid beam) imaging collimators on the face of opposed gamma camera detectors. The combination of these devices in this arrangement permits dual energy imaging to be accomplished either by placement of a rotating slotted filter in the X-ray beam or by electrically pulsing the X-ray tube at two different voltages to generate two energy levels in the transmitted X-ray beam. The gamma camera could be windowed and gated to these two different energies so as to record the images in two separate memory locations, one for each transmission energy. The use of X-ray tube transmission sources in TERI scanner designs using converging collimators is cited as a method of generating transmission images on gamma cameras without isotopic sources for the purpose of transmission CT imaging.

16. A hardware interface buffer for multienergy transmission/emission TERI scanner fan-beam-collimated SPECT gamma camera angle view memory array/next computer interface buffer as shown in Fig. 11

TRANSMISSION/EMISSION REGISTERED IMAGE (TERI) SCANNER

Note: All parts mounted on rotating gantry.



1 Triangular multivaned lead "fan beam" transmitting collimator. The line focus of this collimator is congruent with the line focus of the opposing "fan beam" imaging collimator.

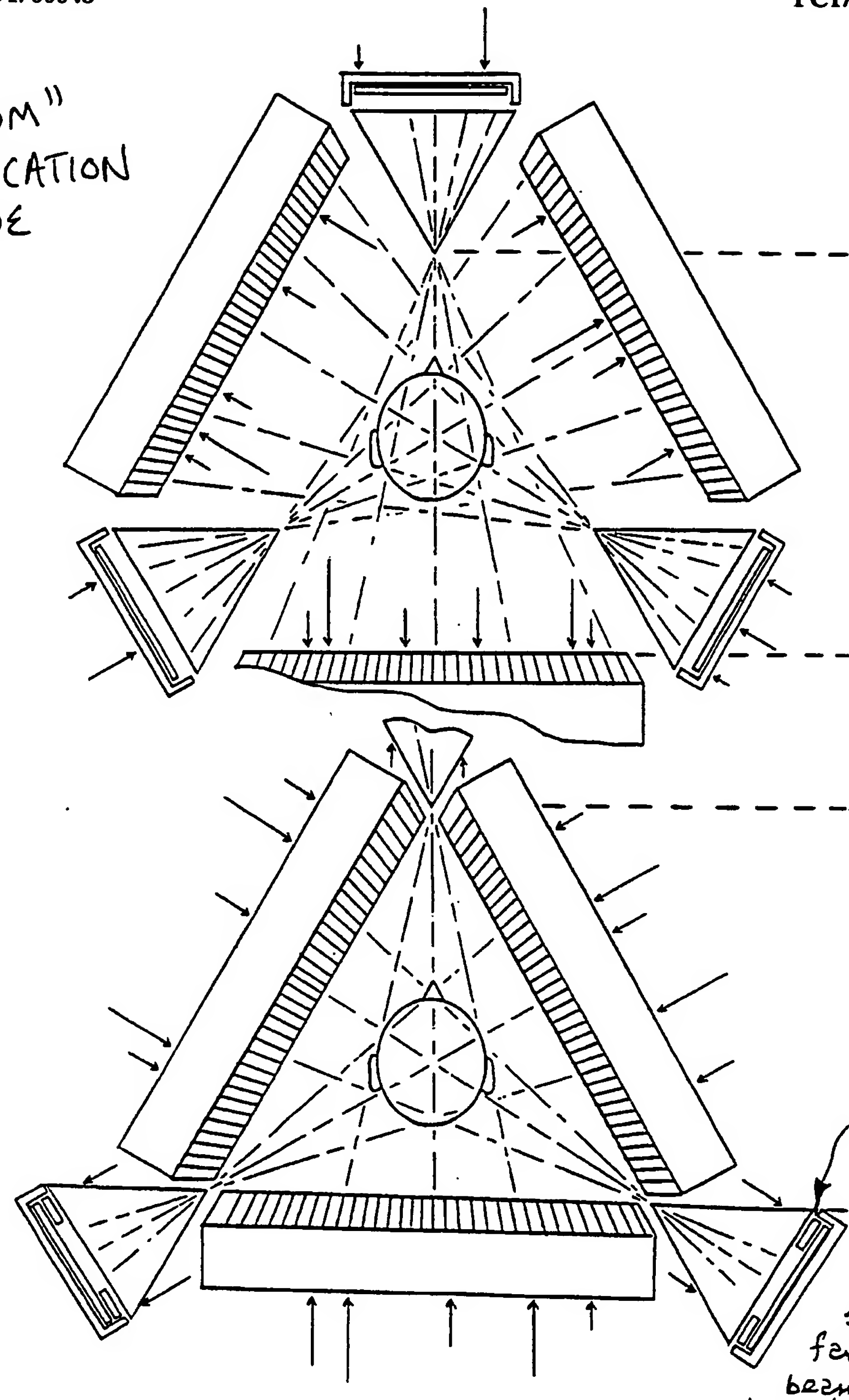
2 Moveable radioactive sheet source attached to servo-control motor to move source into or out of lead box. Multiple sheet sources could be stored in lead box for energy subtraction via serial transmission images, on sequential gantry rotation.

3 Microcast fan beam (line focusing) imaging collimator. Line focus is congruent with the line focus of the opposit "transmitting" collimator.

4 Gamma cameras.

5 Lead shielded box.

"ZOOM" MAGNIFICATION MODE



Note insertion of lead attenuatio shims in "Non-zoom mode to narrow the spread of te fan transmissi beam to ~~the~~ eliminate unnecessa scatter and reduce "noise" coun

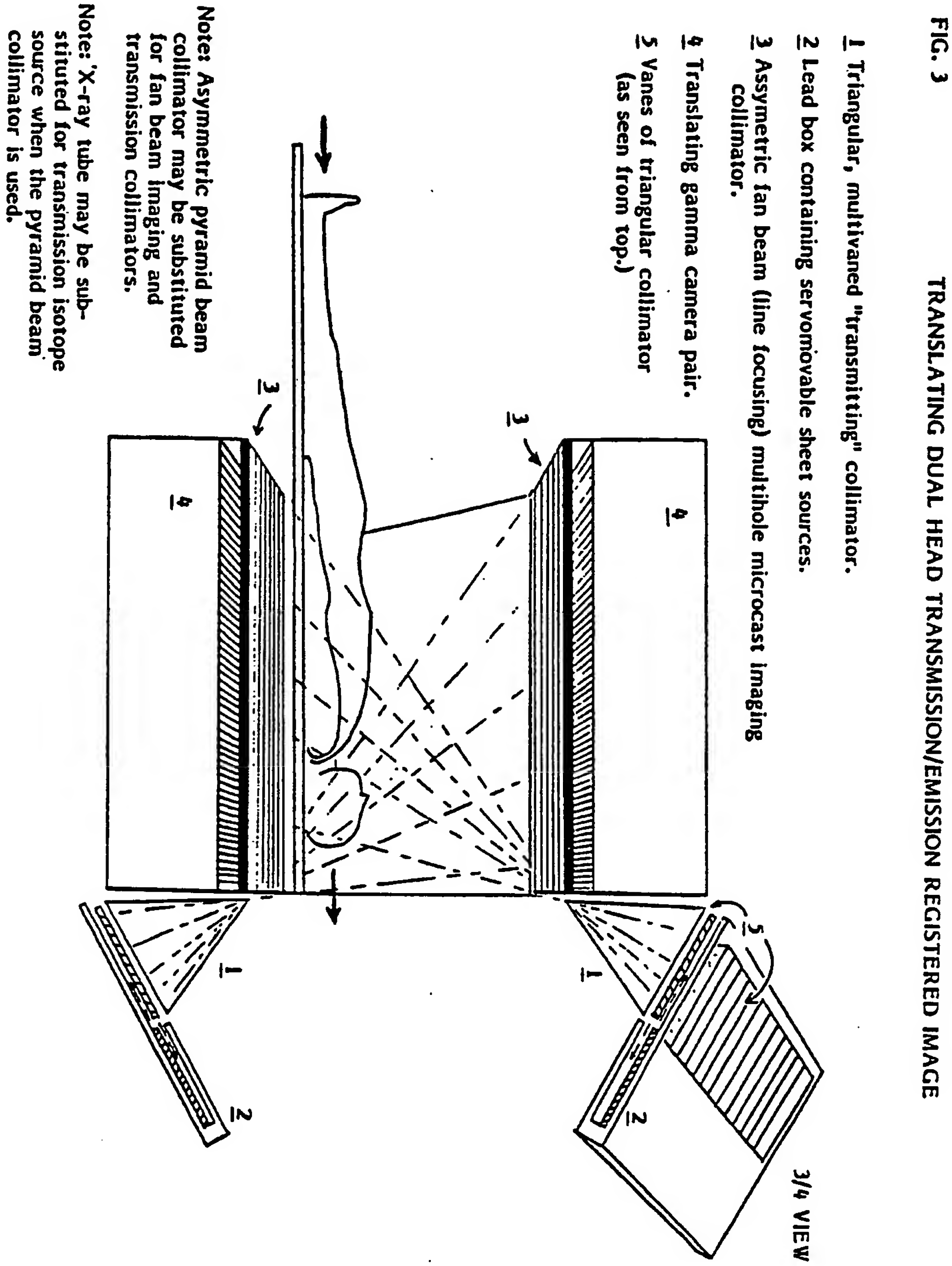
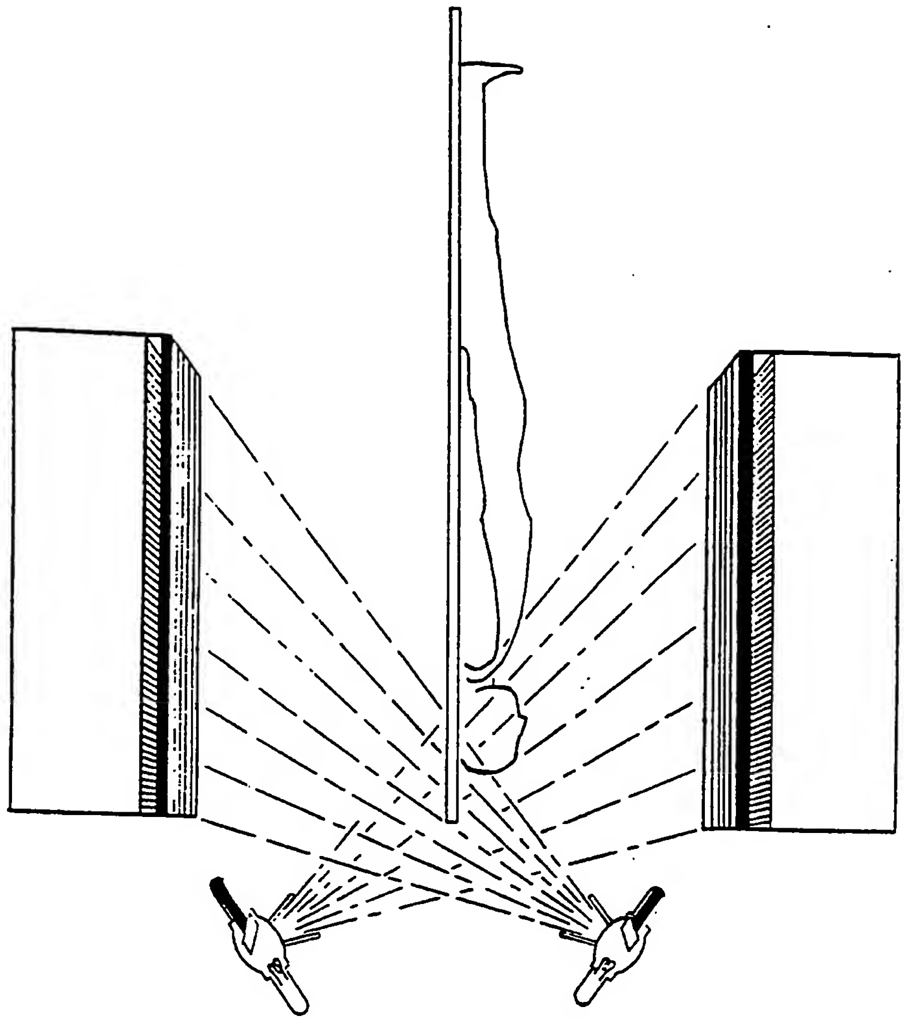
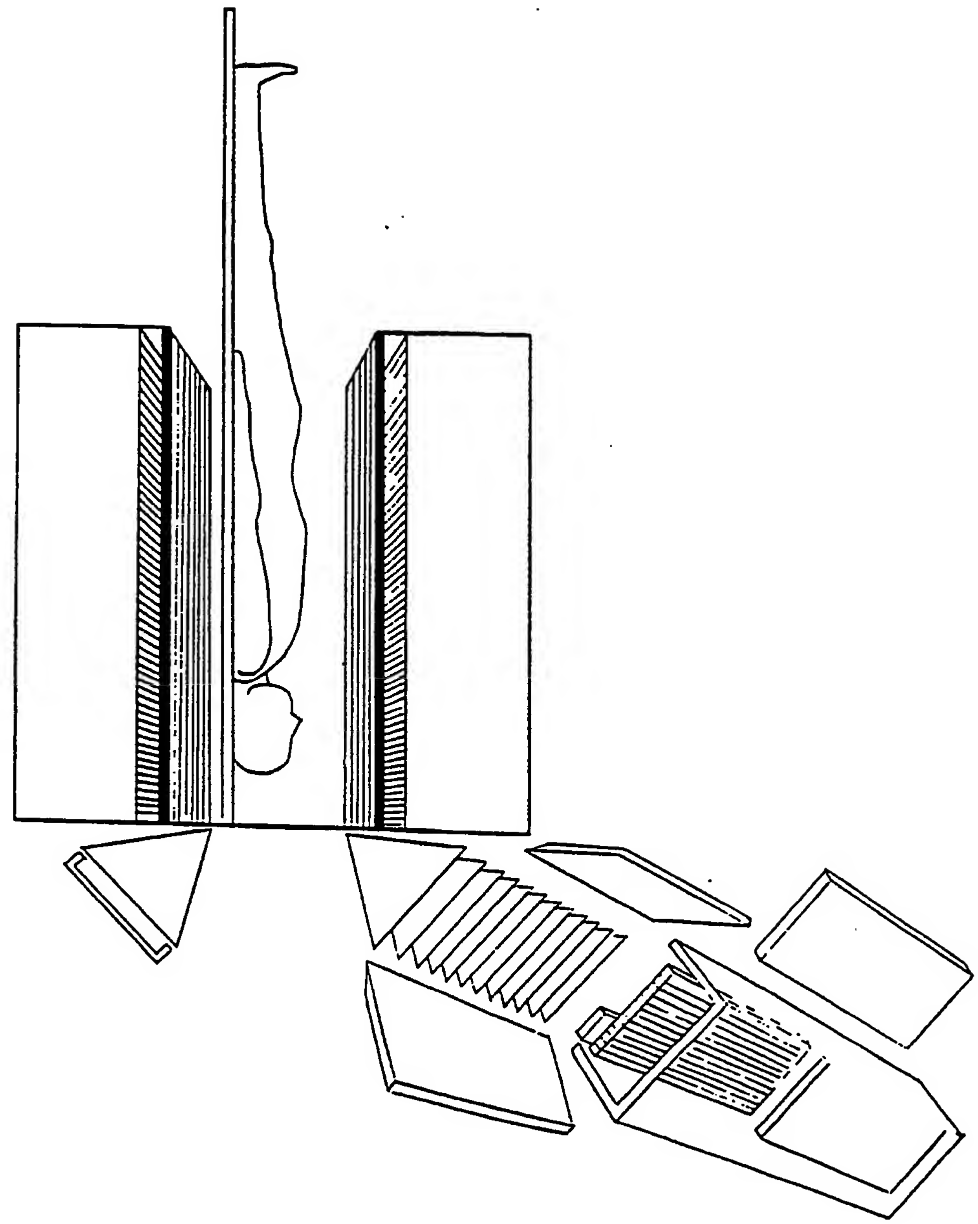
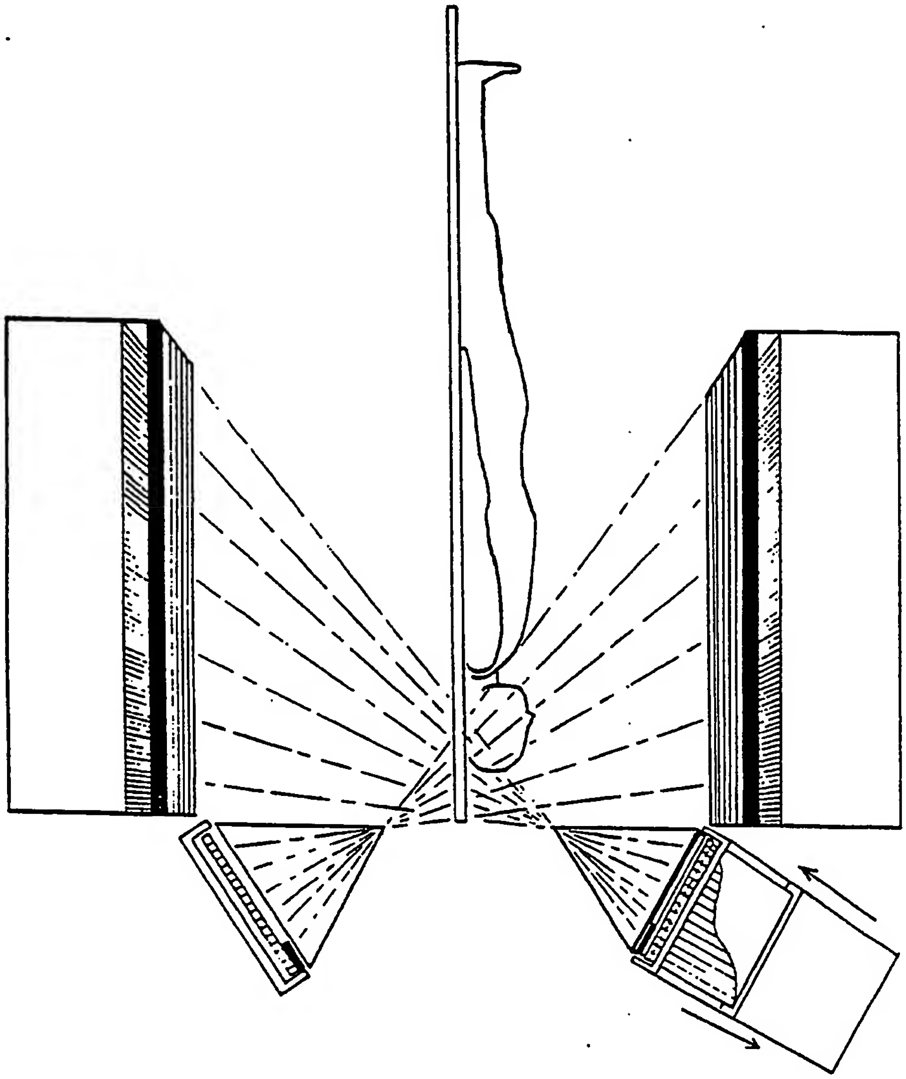
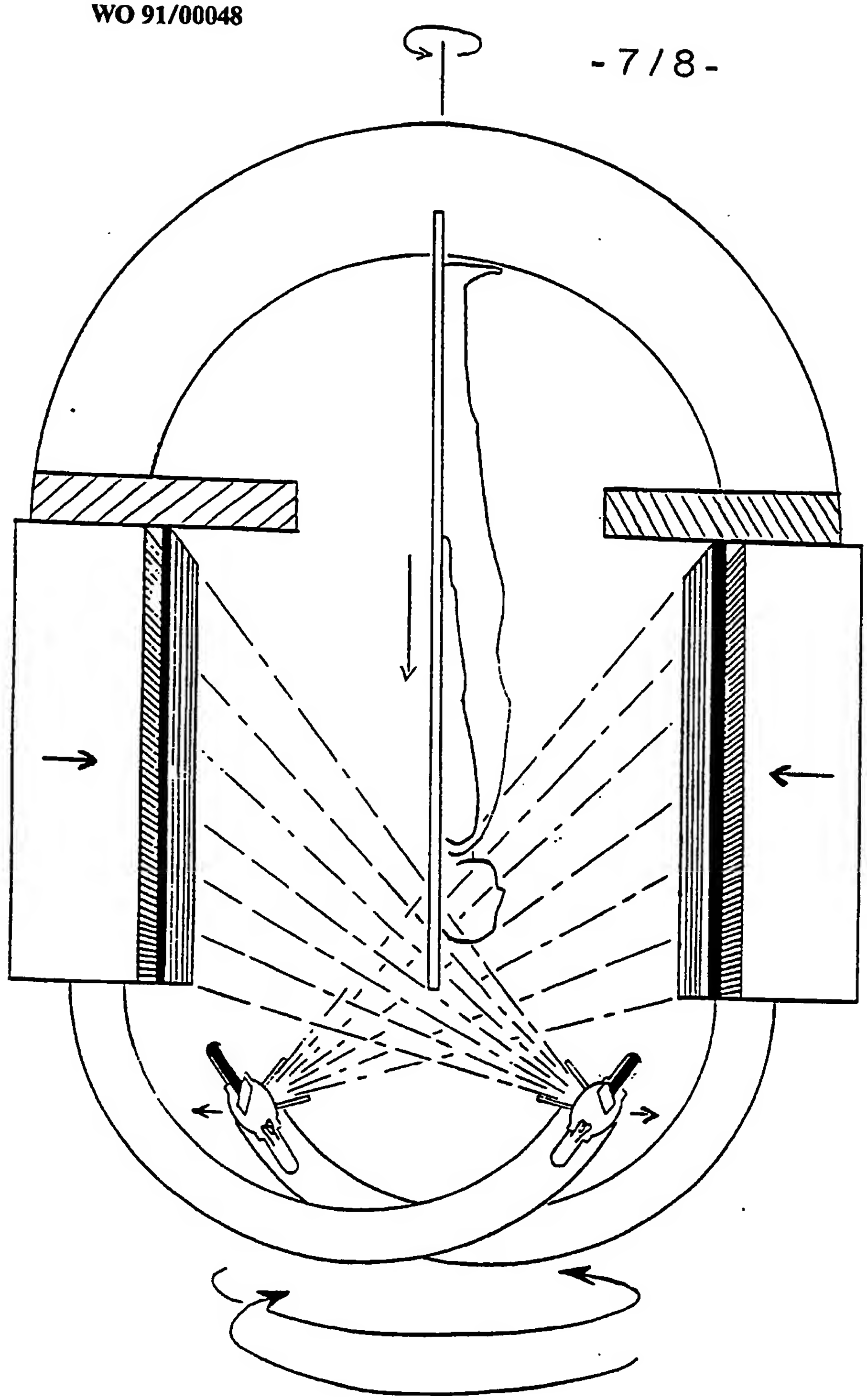


FIG. 3
TRANSLATING DUAL HEAD TRANSMISSION/EMISSION REGISTERED IMAGE
1 Triangular, multivaned "transmitting" collimator.
2 Lead box containing servomovable sheet sources.
3 Assymetric fan beam (line focusing) multihole microcast imaging collimator.
4 Translating gamma camera pair.
5 Vanes of triangular collimator (as seen from top.)
Note: Asymmetric pyramid beam collimator may be substituted for fan beam imaging and transmission collimators.
Note: X-ray tube may be substituted for transmission isotope source when the pyramid beam collimator is used.
3
3
4
4
1
1
2
2
5
3/4 VIEW









3/4 EXPLODED VIEW